INVESTIGATION OF HEART-LUNG FUNCTION IN SMOKERS USING INTEGRATED HEART-LUNG SIGNAL ACQUISITION SYSTEM

A thesis Submitted in partial fulfilment of the requirements for the award of the degree of

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In
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Submitted By:

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CERTIFICATE

This is to certify that the thesis entitled "**INVESTIGATION OF HEART-LUNG FUNCTION IN SMOKERS USING INTEGRATED HEART-LUNG SIGNAL ACQUISITION SYSTEM**" by **SANTOSH KUMAR SAHOO (111BM0545)**, submitted to the National Institute of Technology, Rourkela for the degree of Bachelor of Technology is a record of bonafide research work, carried out by him in the Department of Biotechnology and Medical Engineering under my supervision. I believe that the thesis fulfils part of the requirements for the award of Bachelor of Technology. The results embodied in the thesis have not been submitted for the award of any other degree.

Date-
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Santosh Kumar Sahoo

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ABSTRACT

Integrated heart lung signal acquisition system is a device which helps to acquire ECG and respiratory signals simultaneously and unambiguously. This study deals with the investigation of the functioning of heart and lungs in smokers. The Vernier EKG sensor and Vernier –NTC thermistor were assembled to record the ECG and respiratory signals from 30 subjects. After recording the data were further divided into 2 groups, smokers and non-smokers. R-R interval and the HRV analysis of the ECG signal was done. For the respiratory signal analysis, first respiratory information were extracted from the data and the inspiratory and expiratory volumes were then determined. From the results, it was found that the mean heart rate of the smokers was less than the non-smokers whereas the mean R-R interval was more in case of smokers. From the frequency domain analysis it was observed that there is significant increase in the low frequency (LF) indices in case of smokers and slight increase in the high frequency (HF) indices for non-smokers. Furthermore the LF/HF ratio was higher in case of smokers. The Shannon entropy analysed from the nonlinear indices of HRV was found to be more or less similar for both smokers and non-smokers whereas the Poincare standard descriptors seem to be less for smokers in comparison with the non-smokers. The analysis of the respiratory signals showed that both the inspiratory and expiratory areas under the respective curves were higher for the smokers in comparison to the non-smokers. From all these we can deduce that the sympathetic activity in case of smokers is high as compared to the non-smokers. Thus, the overall study concluded that the integrated data acquisition system is a good methodology to investigate both ECG and respiratory signals. In future, this methodology can be used to find the relation between R-R intervals and breathing intervals to investigate the co-relative function between heart and lungs in various pathological cases.

Keywords: Integrated data acquisition system, ECG, respiratory, R-R interval, HRV

Chapter 1

INTRODUCTION

1. INTRODUCTION

Heartbeat dynamics and respiration dynamics are two of the most studied physiological processes primarily because of their clinical importance and complexity. These two processes are known to influence each other [1]. The study of both respiration dynamics and heartbeat dynamics carry high clinical significance.

1.1 PHYSIOLOGY OF HEARTBEAT AND RESPIRATION

The human heart is considered to be the most vital organ of our body. The functioning mechanism is complex and unique [2]. The heart has its own system of conducting and generating action potentials through a series of complex mechanism. Ionic concentration across the cell membrane changes and the SA (Sino-atrial) node located at the top of right auricle generates impulse my exciting the muscles[3]. This impulse propagates through the right and left auricles with a velocity of 1m/s. The impulse then travels to the junction of the two auricles and ventricles. The AV (auriculo-ventricular) node controls the further propagation of the impulse[4]. A group of fibrous non-excitable cells called Purkinje fibres delays the propagation of the impulse by 0.11s which is further carried to the ventricles by a special conduction system called bundle of His[5]. The Purkinje fibres split into two branches and excite the ventricles simultaneously. The impulse propagates through these fibres at a velocity of 1.5-2.5m/s. The direction of propagation of impulse along the bundle of His starts from apex. Thus, the ventricular contraction starts from the apex which in turn pumps out the blood.

The process by which oxygen is inhaled (inspiration) and provided to the tissues and carbon dioxide is exhaled out (expiration) is known as respiration[6]. Respiration is an involuntary act controlled the autonomic nervous system via the medulla oblongata of the brain[7]. The medulla oblongata senses blood levels of carbon dioxide and triggers respiration at increased carbon dioxide levels. During respiration, air is moved into and out of by changing the volume of the lungs. The volume changes in the lungs are facilitated by contractions of skeletal muscles, namely, the intercostal muscles and the diaphragm[8]. The muscles of the thorax and the movement of the diaphragm create negative and positive pressures for expiration and inspiration respectively. The contraction of diaphragm pulls the lower part of the lungs downward during inspiration and the relaxation of diaphragm, recoil of the lungs, chest wall and abdominal structures compresses the lungs during expiration[9]. In normal individuals no muscle contraction involved in expiration. This process is simply driven by the elastic recoil of the lungs. However, during periods of higher metabolic rate, e.g. during exercise and during voluntary deep expiration

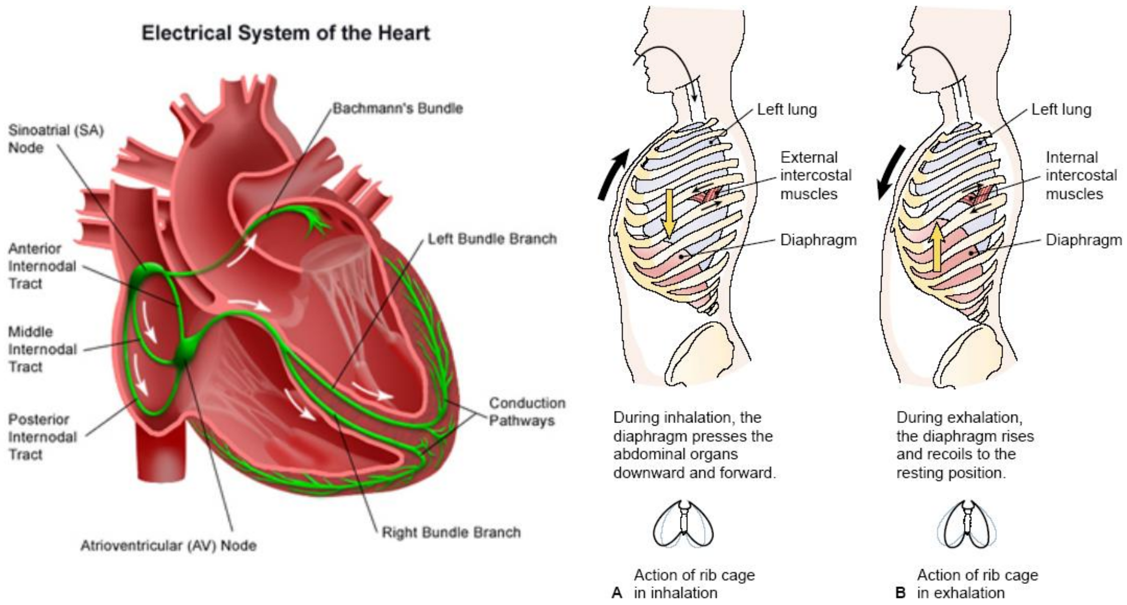


Figure 1: Electrophysiology of heart and pulmonary ventilation.

1.2 IMPORTANCE OF ECG AND RESPIRATORY RATE

1.2.1 Electrocardiogram (ECG)

The recording of the electrical impulse or the electrical activity of heart is known as the electrocardiogram (ECG)[10]. An ECG is a rhythmic, quasi-periodical signal synchronized with the functioning of heart. The normal ECG wave signal is composed of three separate waves: P wave, QRS complex and T wave. The QRS complex consists of three separate waves: Q wave, R wave and S wave[11]. The P wave is caused by the action potential generated by the depolarization of atria prior to contraction. The QRS complex is caused by the potentials generated by the depolarization of ventricles before contraction. The T wave is caused by potentials generated by the repolarization of the ventricles. The repolarization of ventricular muscles occur after 0.25-0.35s of depolarization[12].

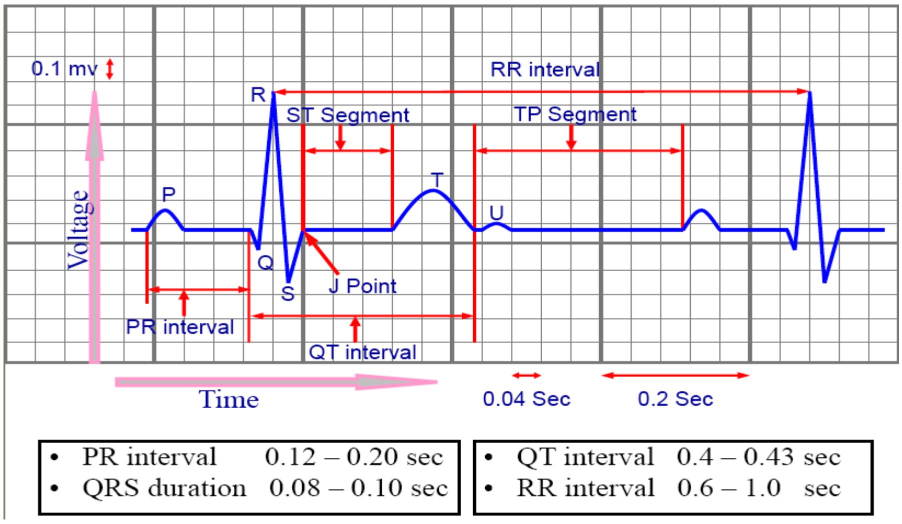


Figure 2: Segmentation of a normal ECG signal.

1.2.2 Respiratory Rate

Breathing is one of the main body functions and its alterations have been proved to be a risk factor in different patient population[13]. The analysis of respiratory parameters provides various clinical information of diagnostic value. There are several methods which have been implemented for the measurement of respiratory rate. So most commonly used methods for measurement of respiratory rate can be categorized into direct methods (e.g. thermistor method, displacement method) and indirect methods (e.g. impendence method)[14].

- <u>Thermistor method:</u> The measurement of air flow through the mouth and nose is considered to be the gold standard which is usually done by temperature measurement of inhaled and expelled air. In this method the difference in temperature between the inspired and expired air is measured. This is a direct method and the temperature difference is detected by placing a thermistor (or thermocouple) in-front of nostrils[15].

- <u>Displacement method:</u> The respiratory cycle is accompanied by changes in thoracic volume. These changes can be detected by means of a displacement sensor that uses a strain gauge[16]. The sensor is enclosed by an elastic band which is tied around the waist. Respiratory efforts result in resistance changes of the strain gauge connected to an arm of a Wheatstone bridge.

- Impedance Pneumography: This is an indirect method used for measurement of respiration rate. This involves external electrodes which are applied on the thorax. These electrodesmeasure respirationrate using the relation between respiratory depth and change in thoracic impedance[17]. A highfrequency current through the electrodes is used and the modulated signal is detected. The change in thoracic impedance is due the respiratory movement

1.2.3 Relationship between HRV and Respiratory rate

Heartbeat and respiration are two vital physiological functions and respiration is considered to be the most important modulator of heart rate as well as source of the short term heart rate variability[18]. The effect of breathing pattern on heart rate changes is called respiratory sinus arrhythmia (RSA) while the inverse is known as cardio-ventilatory coupling[19]. Analysis of ECG with respect to respiration can give important insight into the autonomic control of heart rate as both of these functions are modulated/regulated by fluctuations of the autonomic nervous system (ANS). So far HRV has been used extensively as a non-invasive tool to investigate the autonomic control of the cardiovascular system. HRV signals contain well defined rhythms. Rhythms in the low frequency (LF) range, between 0.04 to 0.15 Hz, are usually considered as markers of sympathetic modulation while the rhythms in high frequency (HF) range, between 0.15 to 0.4 Hz, are regulated by parasympathetic activity[20]. Vagal activity is considered to be the major contributor to the HF component. Respiration, also has a significant effect on the HR oscillations and parasympathetic activity is very closely related to respiratory sinus rhythm. Thus, analysis of respiratory peak can be used as a quantitative measure of vagal control. Respiratory period and tidal volume were shown also to influences RSA as well as the power spectrum of heartbeat interval.

1.3 IMPORTANCE OF HEART LUNG SIGNAL ACQUISITION SYSTEM

There has been radical growth in the field of mobile health monitoring. Also respiration has long been shown to be correlated with certain aspects of the ECG. Various applications require the simultaneous monitoring of respiration and ECG.

However, a subject made uncomfortable by the numerous sensors placed on them is unlikely to react in a natural fashion, possibly compromising the results. Minimising the number of hardware and sensors required to perform these measurements is an important consideration.

Though there has been extensive study and analysis of HRV, ‘respiratory rate variability’ has not been established yet to such extent. It is more difficult to determine the instantaneous respiratory rate unambiguously. Thus, it has been confirmed that there is no such system which can measure the overall integrity and interrelation of both the physiological systems. Keeping that in mind that there is no such system available and the results of this project may help in developing such a system we are trying to correlate the heart lung functions using an integrated signal acquisition system.

1.4 OBJECTIVE

- Signal acquisition:
  To acquire ECG and Respiratory signals from subjects using Heart-Lung signal acquisition system.
- Data analysis:
  - To extract the ECG features: R peaks and R-R interval etc.
  - HRV analysis of time domain, frequency domain and nonlinear indices.
  - Respiratory signal analysis and feature extraction.
- Comparison: The same methodology can be applied for different pathological cases (like asthma) for detection of any difference in heart-lung functioning from normal person.

Chapter 2

LITERATURE REVIEWS

2. LIETRATURE REVIEWS

2.1 MOTIVATION

There have been studies regarding the linear and non-linear correlations between the heart rate variability and respiratory dynamics over short term and long term spectra. Heart and lungs are actually interconnected unit where the heart pumps blood, it passes through the lungs and again through the lungs it enters into the heart. Studies have shown and proved the relationship between HRV and Autonomic Nervous System (ANS). The vagal (one of the major nerves in the ANS) activation is regulated by the expansion and passive recoiling of the lungs [1]. Thus, in other words the HRV is controlled by the rate of inspiration and the rate of expiration. From various studies it has been confirmed that there is no such system which can measure the overall integrity and interrelation of both the physiological systems. Keeping that in mind that there is no such system available and the results of this project may help in developing such a system we are trying to correlate the heart lung functions using an integrated signal acquisition system.

2.2 RESPIRATION ON HRV

It has been observed that the HRV is controlled by the two branches of ANS. However, traditional HRV analysis have failed to isolate the effects of both parasympathetic and sympathetic system [1]. So in this study normal breathing and deep breathing data were analysed and compared using both tHRV analysis and enhanced HRV (eHRV) analysis. Respiratory intervals were used to locate the frequency interval of parasympathetic activity in HRV signal. Also, the eHRV analysis provided accurate detection of parasympathetic and sympathetic control of the heart.

2.3 RESPIRATION ON R-R INTERVAL

Studies have shown that the respiration modulates the ANS activity. Thus, the influence of breathing frequency and tidal volume on the R-R interval power spectra has been observed the R-R interval power at respiratory and low frequencies declined significantly as breathing frequency increased [2] .

2.4 SPECIFIC RESPIRATORY RATES ON HEART RATE AND HRV

It has been observed that different breathing rates have different effects on heart rate variability (HRV) [3]. Data were collected with respiratory rates of 3, 4, 6, 8, 10, 12, and 14 breaths per minute.it was observed the mean heart rates did not differ among these respiratory rates, but respiratory-induced heart rates at lower breathing rates are significantly lower. The results reflected the possible effects of the slow rate of acetylcholine metabolism and the effect of negative resonance at 3 cycles per minute.

2.5 BREATHING RATE ON RELATIONSHIPS BETWEEN R-R INTERVAL AND BLOOD PRESSURE

The relationship between the oscillations in systolic blood pressure and heart period at different frequencies has been studied. It has been observed the respiratory rate is inversely proportional to the R-R (resp.) and systolic blood pressure (resp.) [4]. It is concluded that the relationship between heart rate and systolic blood pressure is independent of the sympathetic effect and is actually frequency dependent.

2.6 HEARTBEAT AND AGE

It is observed that HRV and inter-breath intervals show invariance. This property has been used to etect any effect of disease, gender, age from his heartbeat and respiration signals [5]. An algorithm is used to estimate multifractal spectrum to confirm this relationship. It is observed that the long range dependency in heartbeat dynamics degrades in elderly people whereas there is no significant difference in the heartbeat and respiratory dynamics for the young people.

2.7 RESPIRATORY RATE DERIVED FROM ECG

There have been several techniques used and studied to obtain a respiration signal from an ECG [6]–[8]. It has been observed that measuring the transthoracic impedance using ECG electrodes can give respiratory information of the patients being monitored in hospitals. In this method the respiration signal is obtained directly from the ECG electrodes rather than from ECG signal using special circuitry. The variations in R-R intervals or their reciprocals (caused due to respiratory sinus arrhythmia) can also be used to derive respiration signal from the ECG. ECG filtering, R and RS amplitude based techniques and QRS areas have also been examined and compared with a reference respiratory signal [9]. From this study it has been concluded that the R and RS amplitude based techniques generate the best respiratory signals and have the advantage over ECG filtering and QRS areas. Based on the results and analysis of the derived respiratory signal, new measures of respiratory variability (RV) were defined.

In another study the reliability of the respiratory signal extraction methods from ECG has been investigated [10]. Recording from a piezoelectric breathing sensor was used as reference. It was observed that analysis of the QRS complex coupled with respiratory sinus arrhythmia (RSA) which is also known as interval method showed better results. The ECG mean method also gave results similar to the interval methods.

Also data obtained from the nasal/oral airflow and a Holter ECG during regular day to day conditions has been studied [11]. Respiratory information has been assessed using RSA and EDR methods. It has been observed that in younger people both methods give reasonable result whereas EDR should be preferred in the elderly people.

It has been also studied the beat-to-beat variability of ECG can also be described as coefficients of the principal components [7].The respiratory rate obtained from the principal component analysis was almost accurate when compared to the breathing rate determined by the Fourier analysis.

2.8 RESPIRATORY INFORMATION FROM MIXTURE OF PHYSIOLOGICAL SIGNALS

The beat morphology method [12] which includes QRS complex analysis (area or amplitude) and HRV has also been studied for estimating the respiratory frequency from ECG . It has been observed that combining the results obtained from time and frequency domain gave the most accurate breathing rate when compared to a reference breathing rate data obtained from chest plethysmography. In another study signals from respiratory inductive plethysmography, 3D acceleration and ECG were mixed to derive the breathing rate data [13]. Mixing three kind of signals helped to derive the respiratory rate more accurately. This method also proved to be more reliable over a dynamic range of respiratory rate

It has been also studied that estimation of the respiratory rate from decomposition of signals like ECG, BP and photoplethysmographic (PPG) signals is also helpful where the persons with cardiac arrhythmias are to be studied in correlation with respiratory information [14]. The use of derived Intrinsic Mode Functions (IMF) using Empirical Mode Decomposition (EMD) has efficiently extracted respiratory information from ECG, BP, PPG signals.

2.9 CHANGING RESPIRATORY RATE ON HRV(PORTRAYED BY DESCRIPTORS OF POINCARÉ PLOT ANALYSIS)

Poincaré plot analysis (PPA) is a nonlinear method of HRV with two standard descriptors, SD1 and SD2. These descriptors of PPA are sensitive enough to detect respiration induced changes in HRV and are correlated with other HRV measures and BRS. The area (S) of the ellipse corresponding to the total HRV and the ratio of SD2/SD1 reflecting sympatho-vagal balance can also be derived from this plot [15]. It has been observed that the increasing respiratory rate caused a significant reduction of SD2/SD1 and S.

Chapter 3

MATERIALS & METHODS

3. MATERIALS AND METHODS

3.1 VOLUTEERS

The study was conducted on 30 volunteers aged between 19-26 years. Prior to the experimentation the volunteers were informed about experimental details. A written questionnaire about the background information, medial history, smoking habits etc. (see ANNEXURE I). of the volunteers was taken before the commencement of the experiment. The volunteers were suggested to rest on wooden chairs/table in sleeping position to minimize any disturbance. Volunteers were asked not to touch ground throughout the proceedings of the experiment. All the metallic and electronic gadgets (e.g. rings, bracelets, mobile phones etc.) in contact with their body were removed during the test. The ECG and respiratory signals were recorded for 7 minutes (the 1st minute and last minute of each data were filtered later on so that a stable 5 minutes data can be used for analysis). The summary of the participating volunteers is shown in the following table.

Table.1: Information about the volunteers.

CATEGORY	MEAN AGE
NORMAL	21.35± 1.75
ATHLETES	20.685±1.36
SMOKERS	22.18±1.43
MALE	21.47±1.28
FEMALE	21.5±0.5

3.2 MATERIALS

3.2.1 CUSTOM MADE HEART-LUNG CORRELATION INSTRUMENT

The instrument is fully designed and developed in our lab. It consisted of four parts: ECG sensor to measure the heart rate, a thermistor, a prototype and a DAQ.

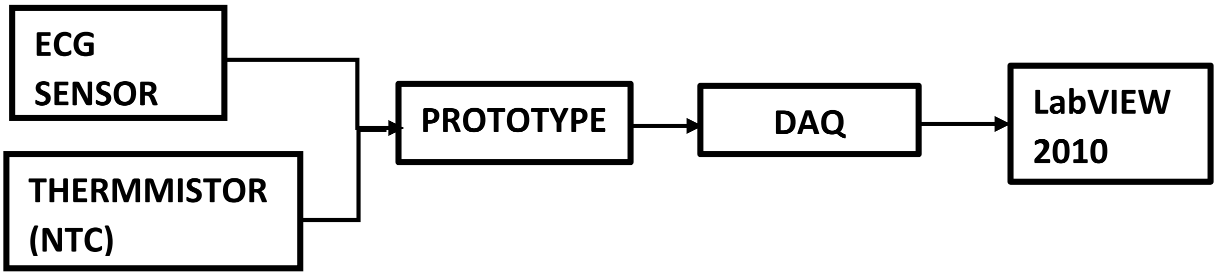


Figure 3: Block diagram representation of the experiment

A) The ECG circuitry consists of 3 bio-potential electrodes which provides electrical contact with the human body and convert these signals into electrical signals. EKG-BTA (*Vernier Software & Technology, Beaverton, OR, USA*) was used as bio-potential amplifier.

Figure 4: EKG-BTA sensor.

[Specifications of EKG-BTA:

Offset: ~1.00 V (±0.3 V)

Gain: 1 mV body potential / 1 V sensor output**]**

B) The Respiratory signal acquisition circuit consists of a thermistor (*NTC, Vernier Software & Technology, Beaverton, OR, USA*) connected to a mask in front of nostrils.

C) The prototype circuit collects the data form both ECG sensors and the thermistor and then transfers them into the DAQ.

D) Both the signals received from the prototype were interfaced with the laptop using a USB data acquisition system (**NI USB-6009**, *National Instruments*, *Austin*, *TX* ,*USA*). LabVIEW 2010 software was used for interfacing the DAQ with the laptop.

The customized system was regarded as Heart-Lung correlation system. The whole setup has been shown in figure 3. The recording was done for 7 minutes and the recorded data was saved as LabVIEW measurement file (.lvm file)

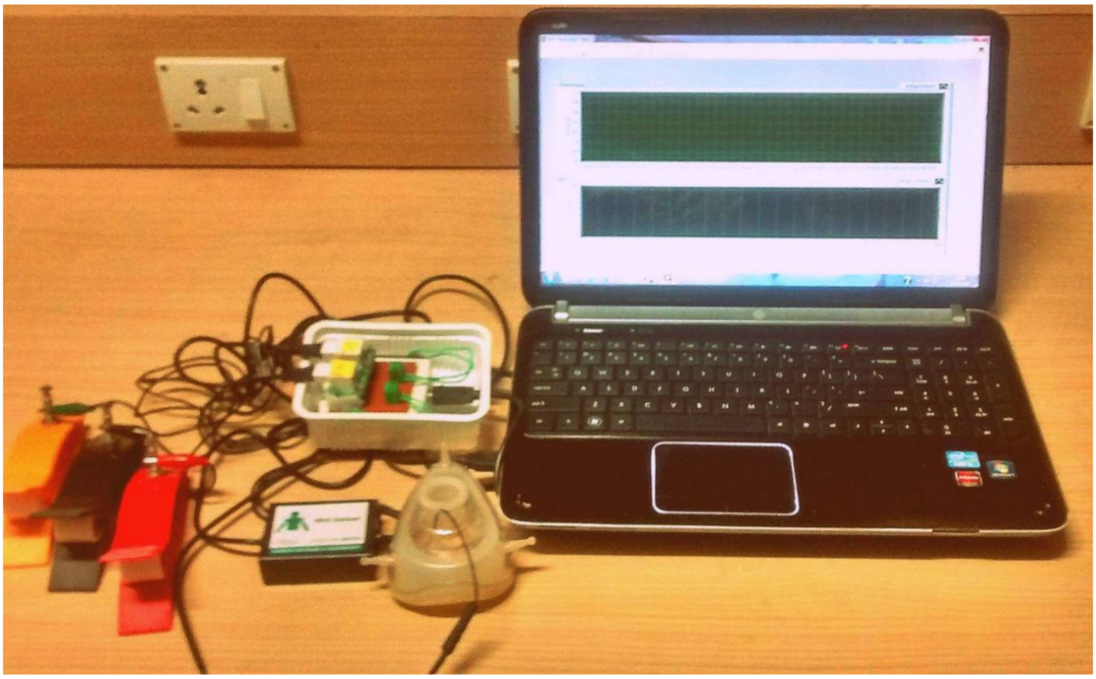


Figure 5: Experimental setup

3.3 METHODS

3.3.1 SIGNAL ACQUISITION

The ECG is acquired by placing an array of electrodes at specific locations on the body surface. Conventionally, electrodes are placed on each arm and leg and six electrodes are placed at different locations on chest. In this experiment standard limb leads were used. The positive electrode is placed on the left arm and the negative electrode on the right arm, measuring the potential difference between them. The difference of the potential measured between two electrodes is with reference to a third electrode which is placed on the right wrist. The ECG measured by the leads is a time variant single dimensional component of vector.

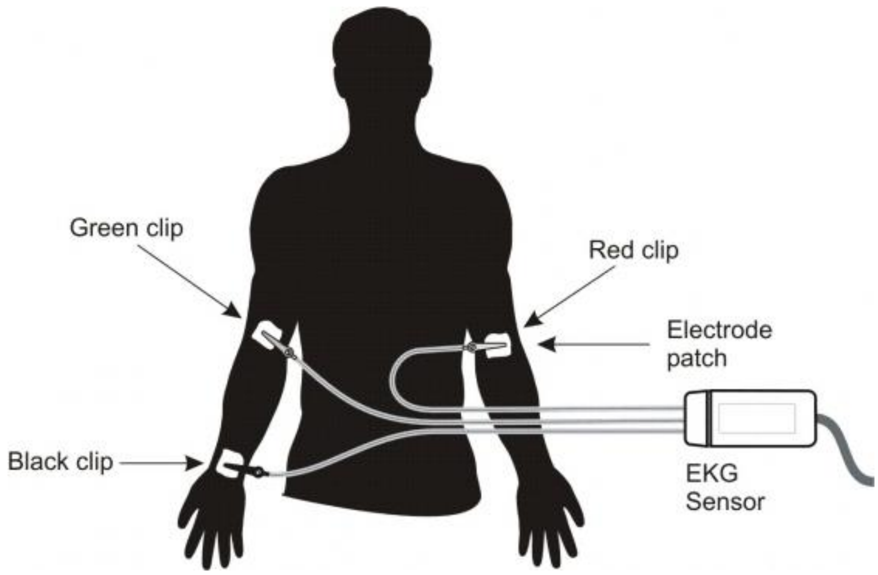


Figure 6: Schematic representation of ECG signal acquisition.

The respiration rate measured in this experiment was carried out by thermistor method [16]. The inhaled air gets warm when it passes through the lungs and the respiratory tract. Hence, there is difference in the temperature between the inspired and expired air. This temperature difference can be detected by a thermistor placed in front of the nostrils.

The signals were interfaced with the laptop using the DAQ and recorded using signal processing tool in LabVIEW 2010 as shown in figure 7 and figure 8.

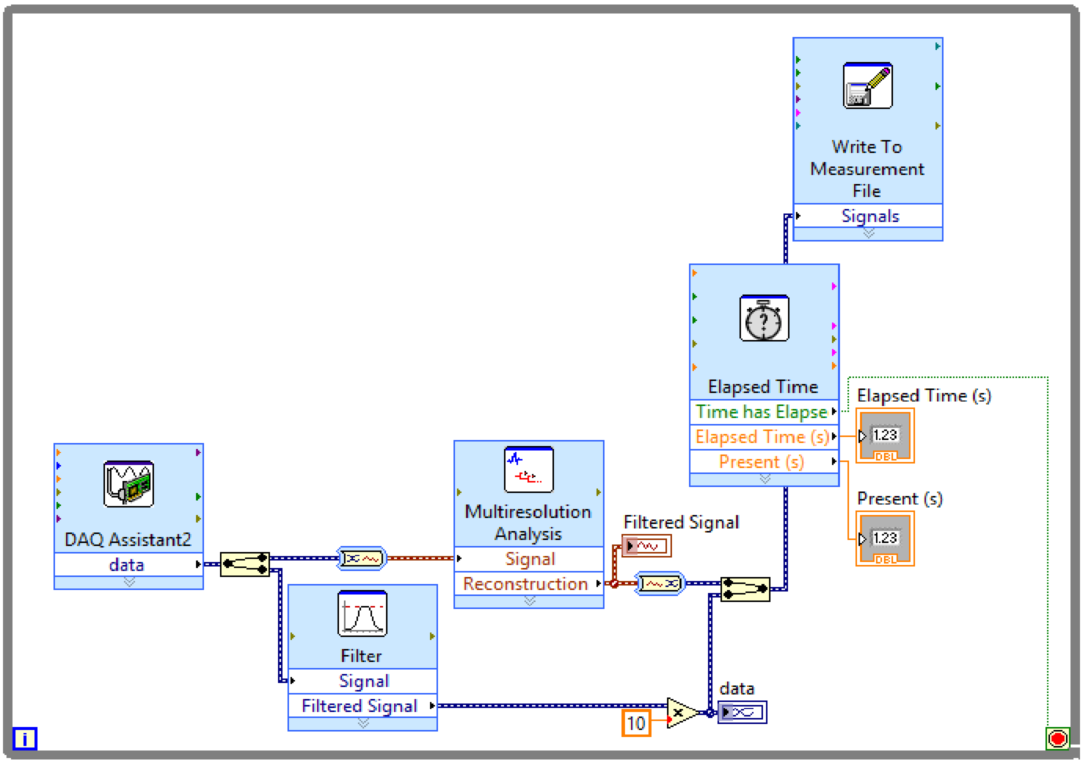


Figure 7: Block diagram in LabVIEW for the experiment.

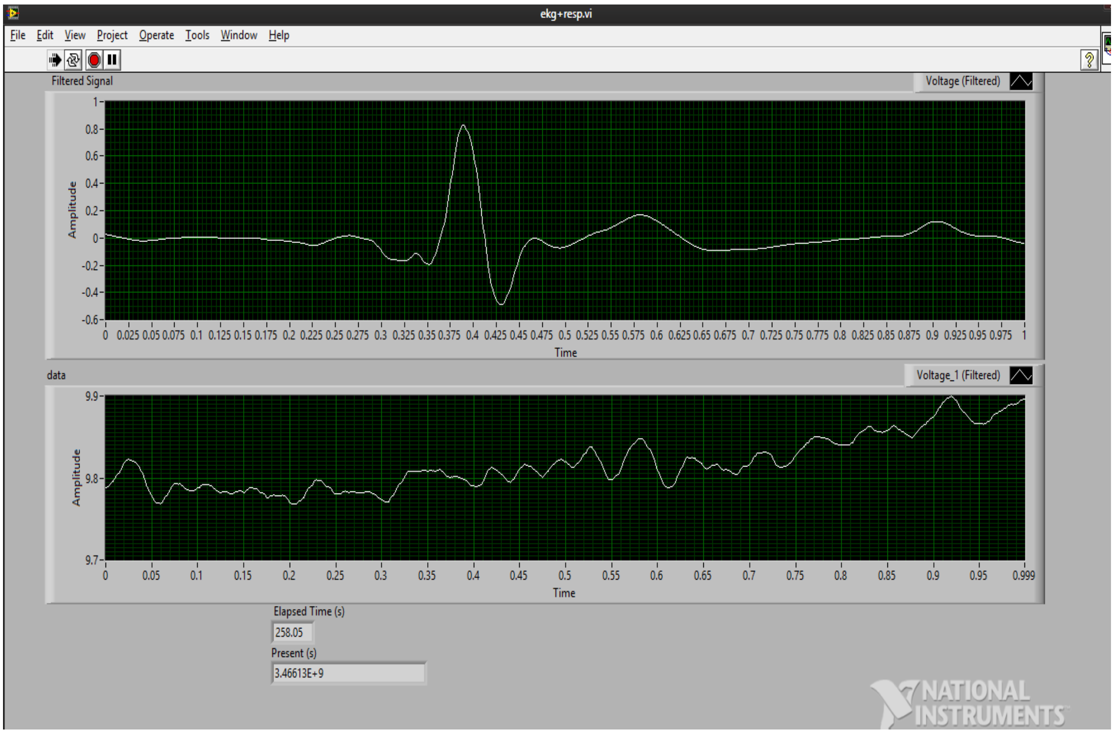


Figure 8: Front panel window showing the ECG and Respiratory rate signals.

3.3.2 R-R INTERVAL ANALYSIS

The R-R interval analysis of the ECG is done manually in Origin 9 Pro. First all the samples were filtered to reduce noise using FFT and then data from 60-360 sec (5mins stable data) were then plotted in Origin 9 Pro. The peak analyser tool is then used to find the R-R peaks. The baseline selection and threshold limit is set manually for each sample respectively.

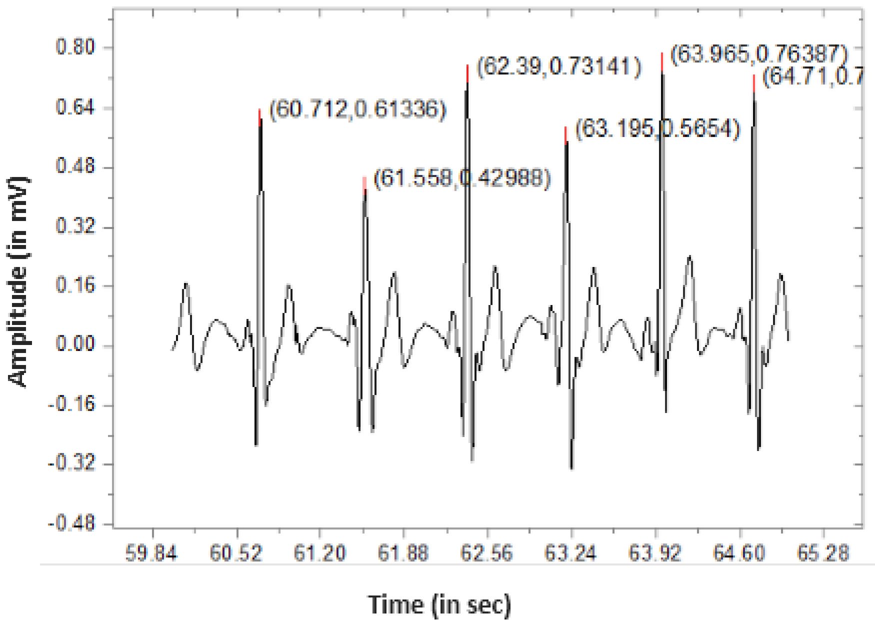


Figure 9: Detection of R-R peaks.

3.3.3 HRV ANALYSIS

Heart Rate Variability provide a non-invasive assessment of various cardio-vascular control mechanisms. The analysis of various linear measures of HRV (time domain, frequency domain indices) and nonlinear indices is done by using NI Biomedical start-up kit 3.0.

3.3.3.1 Time domain indices:

The time domain indices can be derived from direct measurements of the R-R intervals or from the differences between R-R intervals.

- <u>Mean RR</u>: It is the average of the R-R intervals over the entire sample period.

- SDNN: It is the standard deviation of Normal to Normal intervals i.e., the square root of variance, which are often calculated over a period of 24 hrs.
- RMSSD: It is the square root of the mean squared difference of successive Normal to Normal intervals.
- NN50: It is the number of interval differences of successive Normal to Normal intervals which have a difference of greater than 50ms.
- pNN50: It is the proportion derived by dividing NN50 by total number of NN intervals.

3.3.3.2 Frequency domain indices:

The frequency domain indices in HRV analysis provides information about the total variability as a function of frequency. These analysis is then used to estimate result in power spectrum. The spectral analysis of RR intervals gives result in three separate bands:

- VLF: Very Low Frequency band is located in the less than 0.04 Hz region.
- LF: The Low Frequency band is located in the range of 0.04-0.15 Hz range. This basically derives from regulation of blood pressure.
- HF: High Frequency has a range from 0.15-0.50 Hz. This reflect momentary influences of respiratory variables on the heart rate.

The LF/HF ratio is considered to be a convenient index of sympatho-vagal interaction.

3.3.3.3 Nonlinear indices:

In addition to the analysis of the linear indices various studies have shown the assessment of HRV with nonlinear measures may give additional and different information. The analysis of the nonlinear measures involve reconstruction of state space from the sequences of heartbeat periods (time duration between successive R waves). This reconstructed dynamics is later quantified. The analysis also involves fractal analysis.

- Shannon Entropy: It quantifies the repetition of patterns in that signal. Reduce the amount of information by transforming the original time series into set of symbols.
- Approximate Entropy (ApEn): It is approximately equal to the negative average natural logarithm of the conditional probability that two sequences that are similar for ‘m’ data points remain similar within a tolerance ‘r’ at the next point and sample entropy.
- Detrended Fluctuation Analysis (DFA): It refers to the analysis of the series of R-R intervals which are summed cumulatively and then segmented. Then the degree of dispersion of the cumulated time series away from its linear trend is measured. The rate at which total dispersion varies proportional to the segments is measured as a slope on a log-log plot.
- Poincare plot: Poincaré HRV plot is a graph in which each RR interval is plotted against next RR interval with two standard descriptors, SD1 and SD2. SD1: dispersion (standard deviation) of point perpendicular to the axis of line of identity. SD2: dispersion (standard deviation) of points along the axis of line of identity. Poincaré plot analysis is easier and more sensitive at evaluating the sympathovagal balance and observing the beat-to-beat HRV.

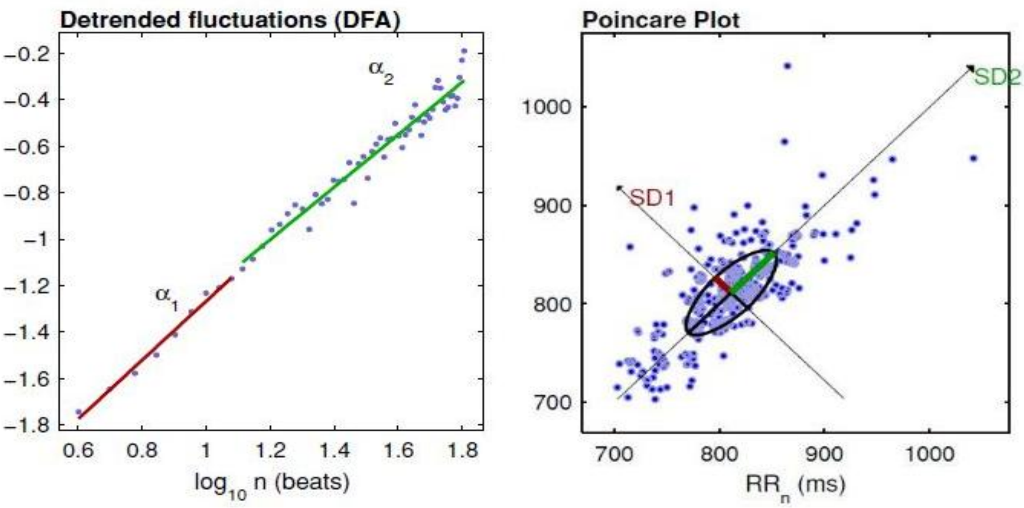


Figure 10: log-log plot for DFA and Poincare plot.

3.3.4 RESPIRATORY SIGNAL ANALYSIS

- These respiratory signals contain external noise. So in order to minimise the noise we have to do filtration and smoothening of the signal. The smoothening of each of the respiratory data was done manually in the Origin 9 Pro (*OriginLab Corporation, USA*) using FFT method.

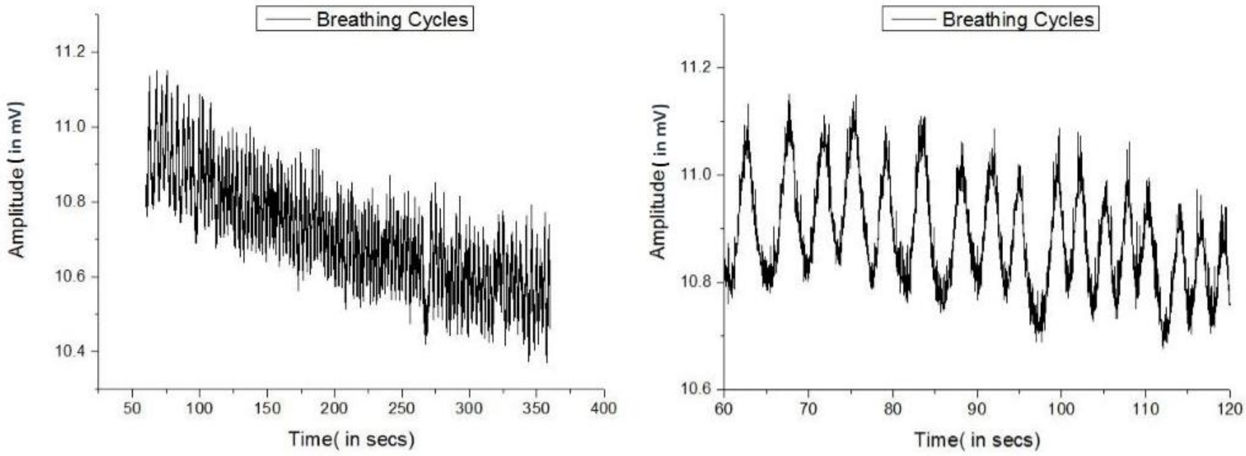


Figure 11: Respiratory signals with noise.

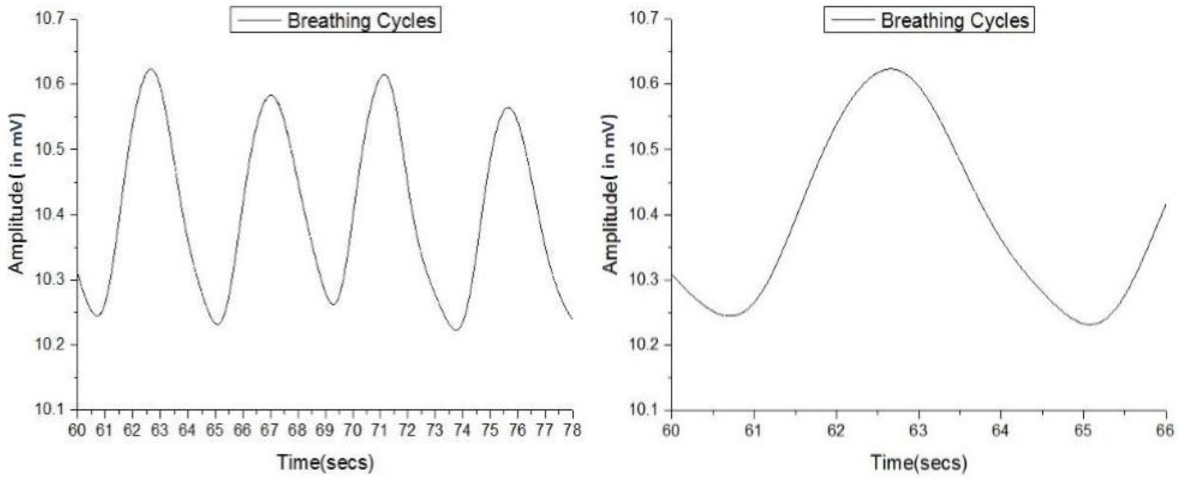


Figure 12: Respiratory signals after noise reduction and smoothening.

- After smoothening and noise filtration, inspiratory and expiratory peeks were detected using Peak Analyser window in Origin 9 Pro. This was done manually using the peak analyser tool in Origin 9 Pro.
- After the peak detection, the inspiratory and expiratory volume calculation was done. This was done by calculating the area under curve for inspiratory portion and expiratory portion respectively using the 'graph integration' tool in Origin 9 Pro as shown in figure 13(see ANNEXURE II).

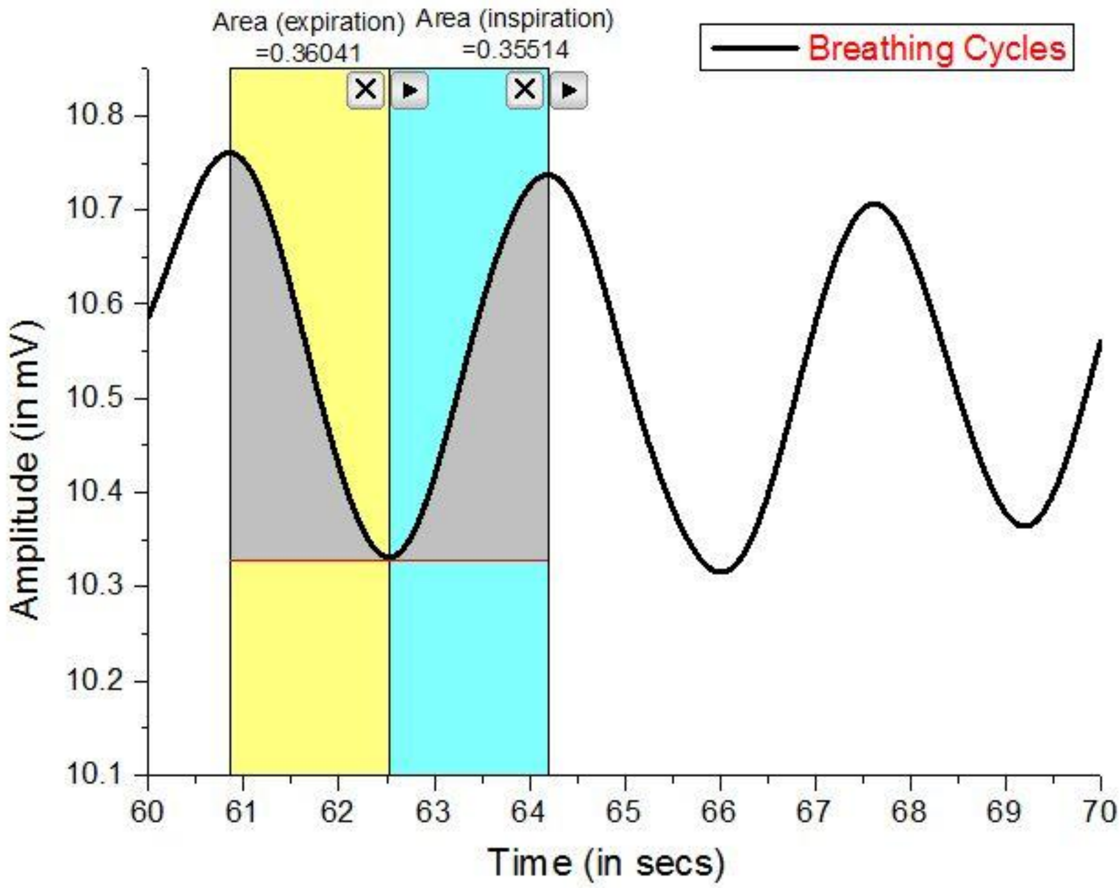


Figure 13: Inspiratory and Expiratory area calculation.

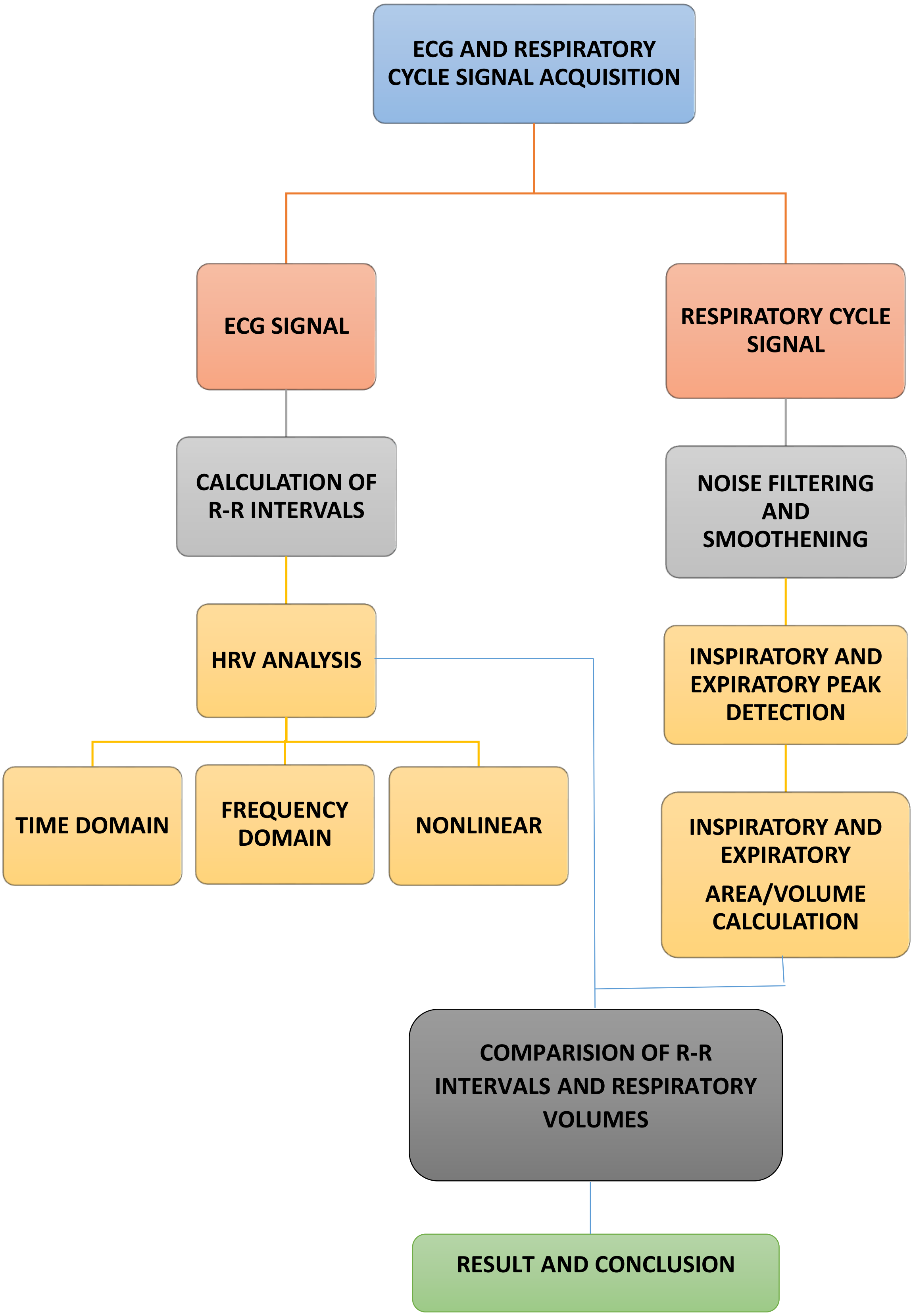


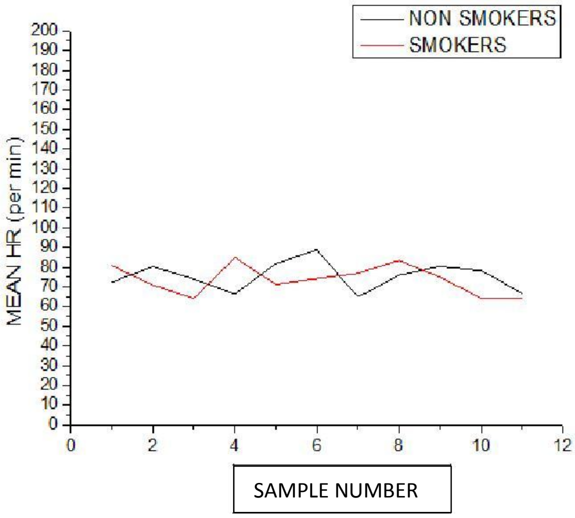
Figure 14: Detailed work plan for the experiment.

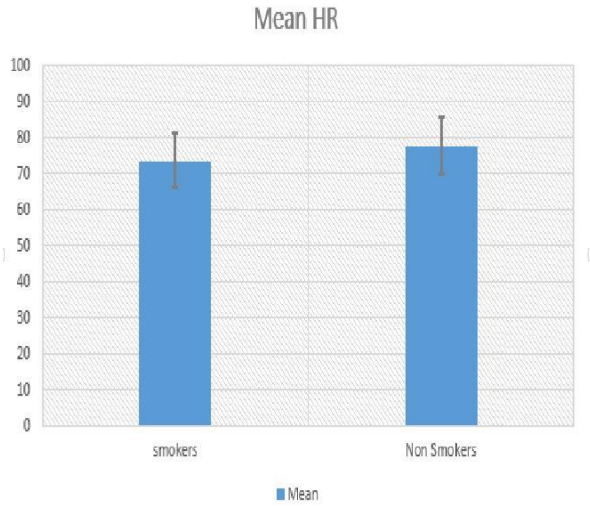
Chapter 4

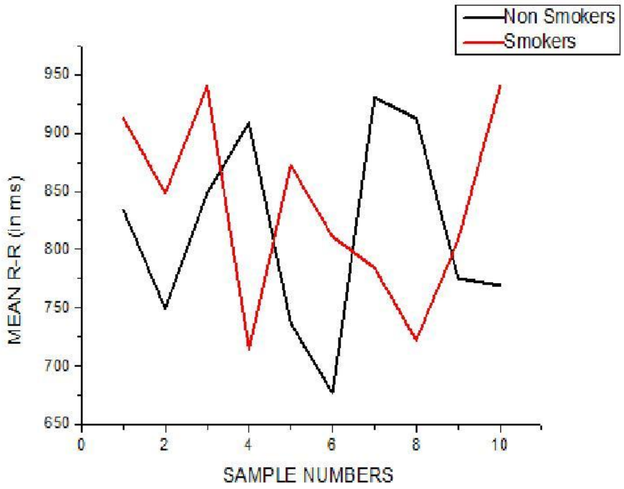
RESULTS
&
DISCUSSIONS

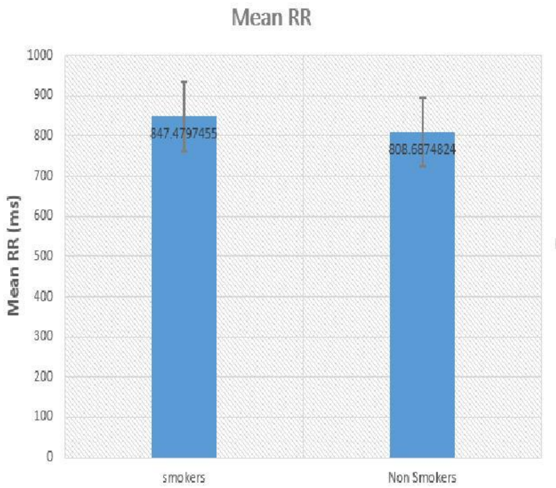
4.1 HRV ANALYSIS

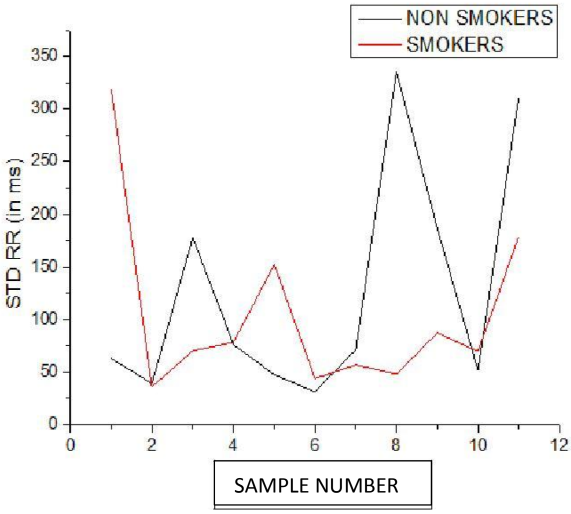
4.1.1 TIME DOMAIN RESULTS

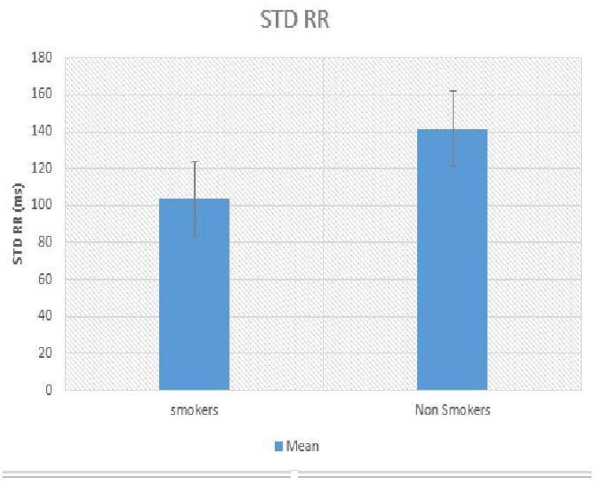












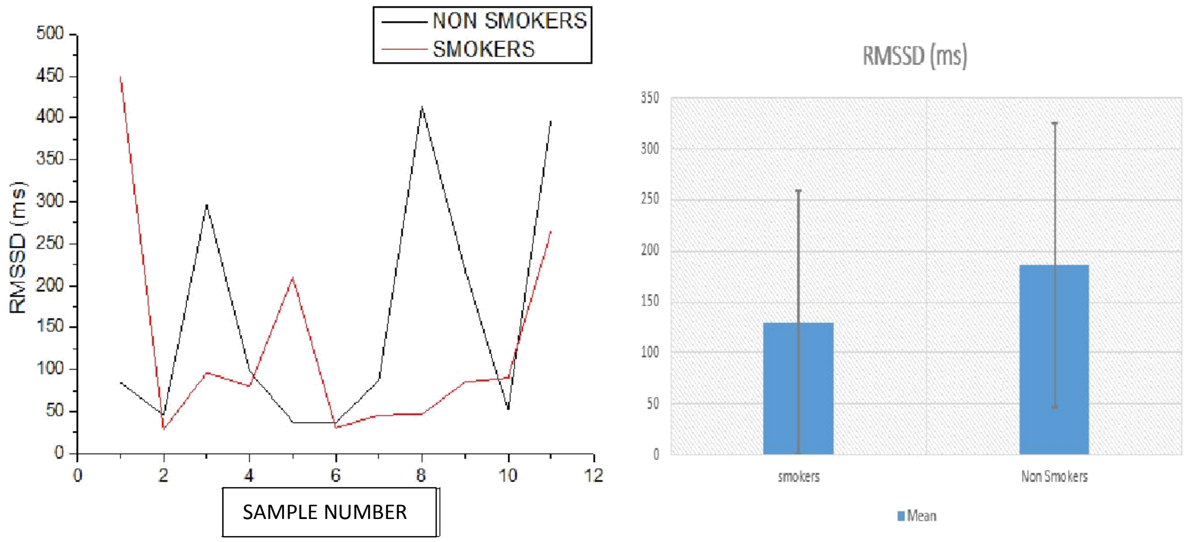


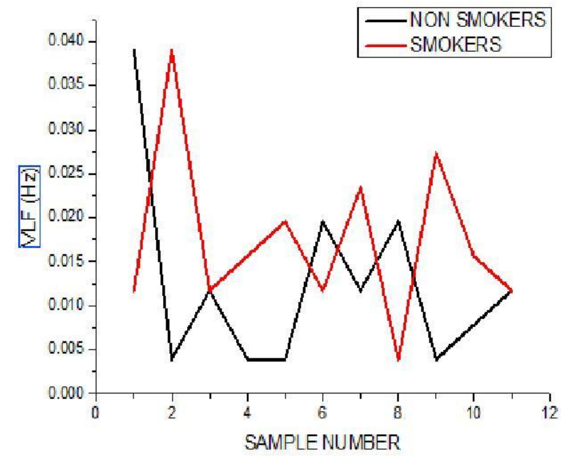
Figure 15: Graphs plotted for Time domain indices.

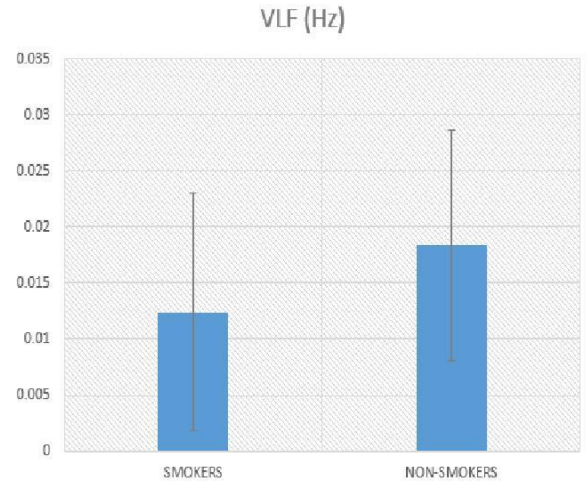
Table 2. : Data obtained from Time Domain analysis.

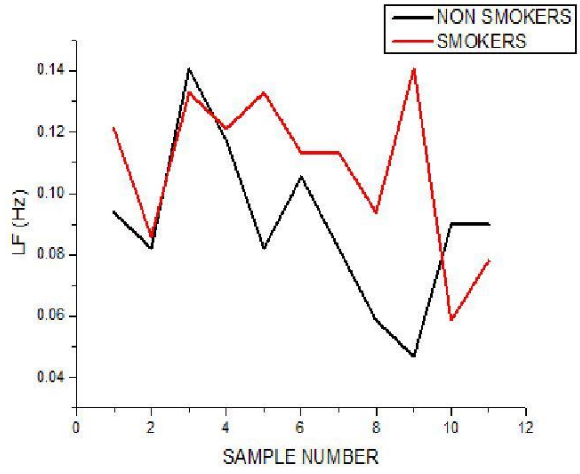
TIME DOMAIN ANALYSIS

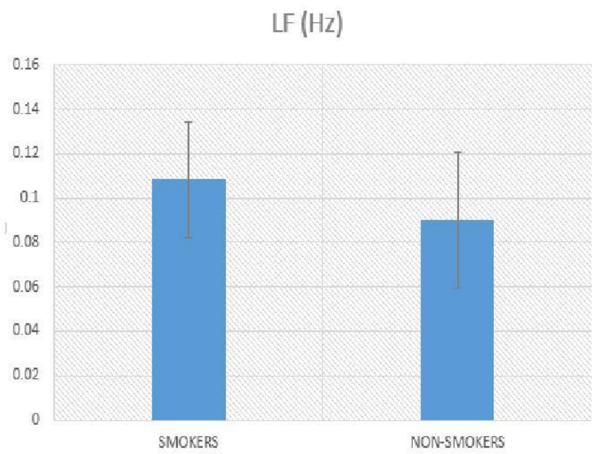
CATEGORY	Mean RR (ms)	STD RR (ms)	Mean HR (1/min)	RMSSD (ms)	NN50 (count)	pNN50 (%)	RR tri index	TINN (ms)
NON-SMOKERS								
Mean	808.6875	141.782	77.80504	185.8198	130.3529	40.362994	8.5944	341.8
Standard Deviation	85.3495	103.7467	8.133753	138.6857	69.57545	21.539578	3.4979	120.5
SMOKERS								
Mean	847.4797	103.5222	73.6477	129.8005	146.6364	48.141073	11.319	326.8
Standard Deviation	87.64406	84.35086	7.517024	109.6523	87.06581	29.212277	5.239	168.2

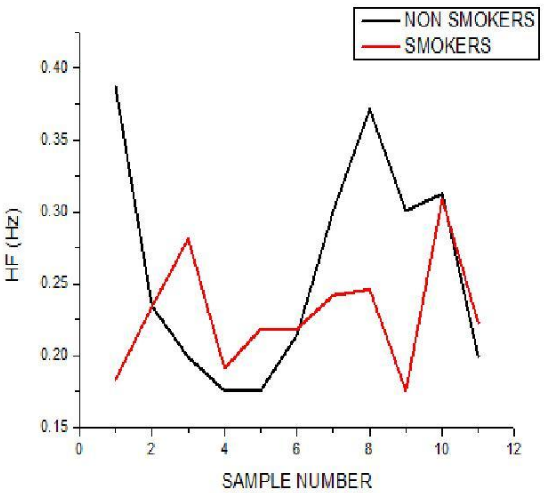
4.1.2 FREQUENCY DOMAIN RESULTS

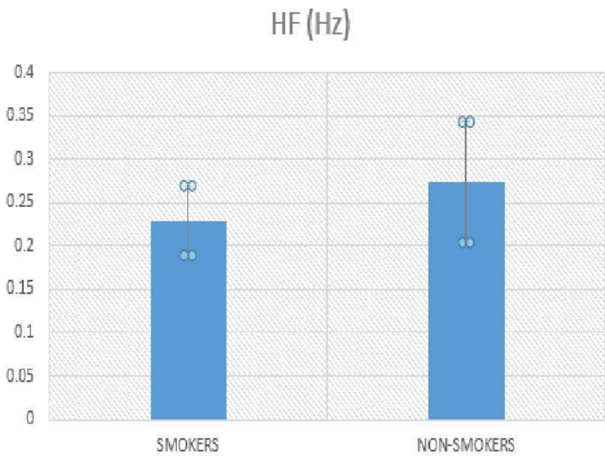


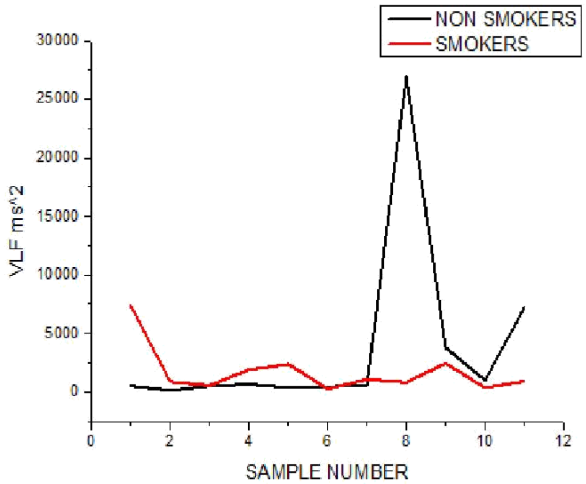


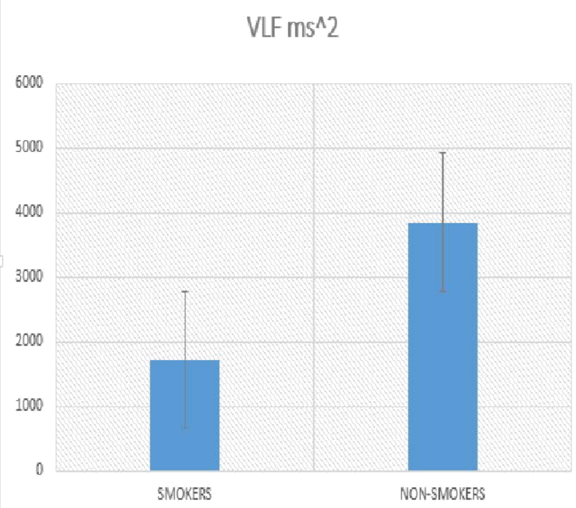


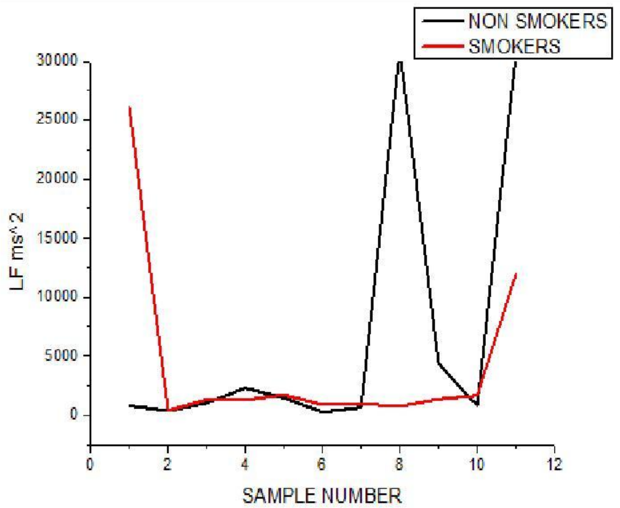


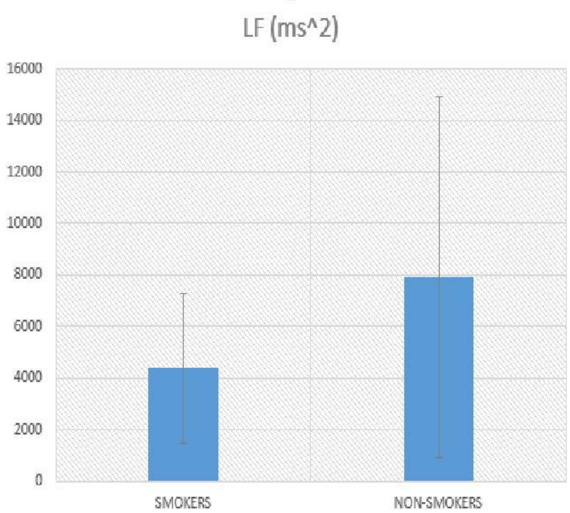


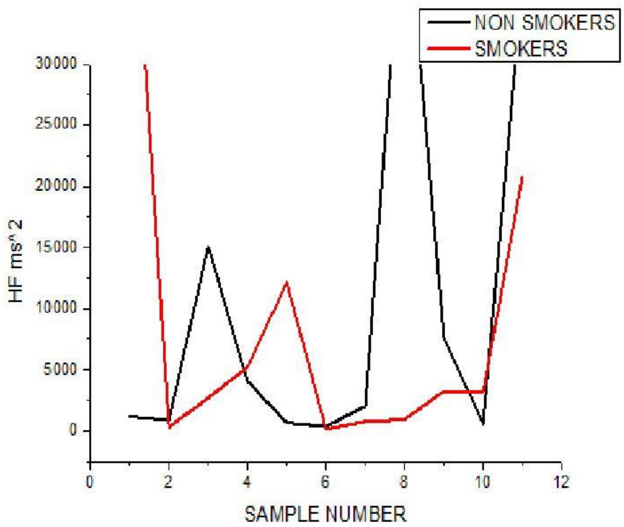


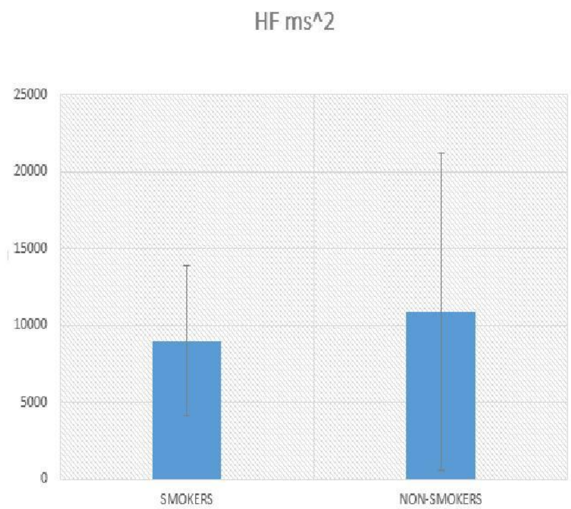

NON SMOKERS
SMOKERS
VLF ms^2
SAMPLE NUMBER
30000
25000
20000
15000
10000
5000
0
0
2
4
6
8
10
12


VLF ms^2
6000
5000
4000
3000
2000
1000
0
SMOKERS
NON-SMOKERS


NON SMOKERS
SMOKERS
LF ms^2
SAMPLE NUMBER
30000
25000
20000
15000
10000
5000
0
0
2
4
6
8
10
12


LF (ms^2)
16000
14000
12000
10000
8000
6000
4000
2000
0
SMOKERS
NON-SMOKERS


NON SMOKERS
SMOKERS
HF ms^2
SAMPLE NUMBER
30000
25000
20000
15000
10000
5000
0
0
2
4
6
8
10
12


HF ms^2
25000
20000
15000
10000
5000
0
SMOKERS
NON-SMOKERS

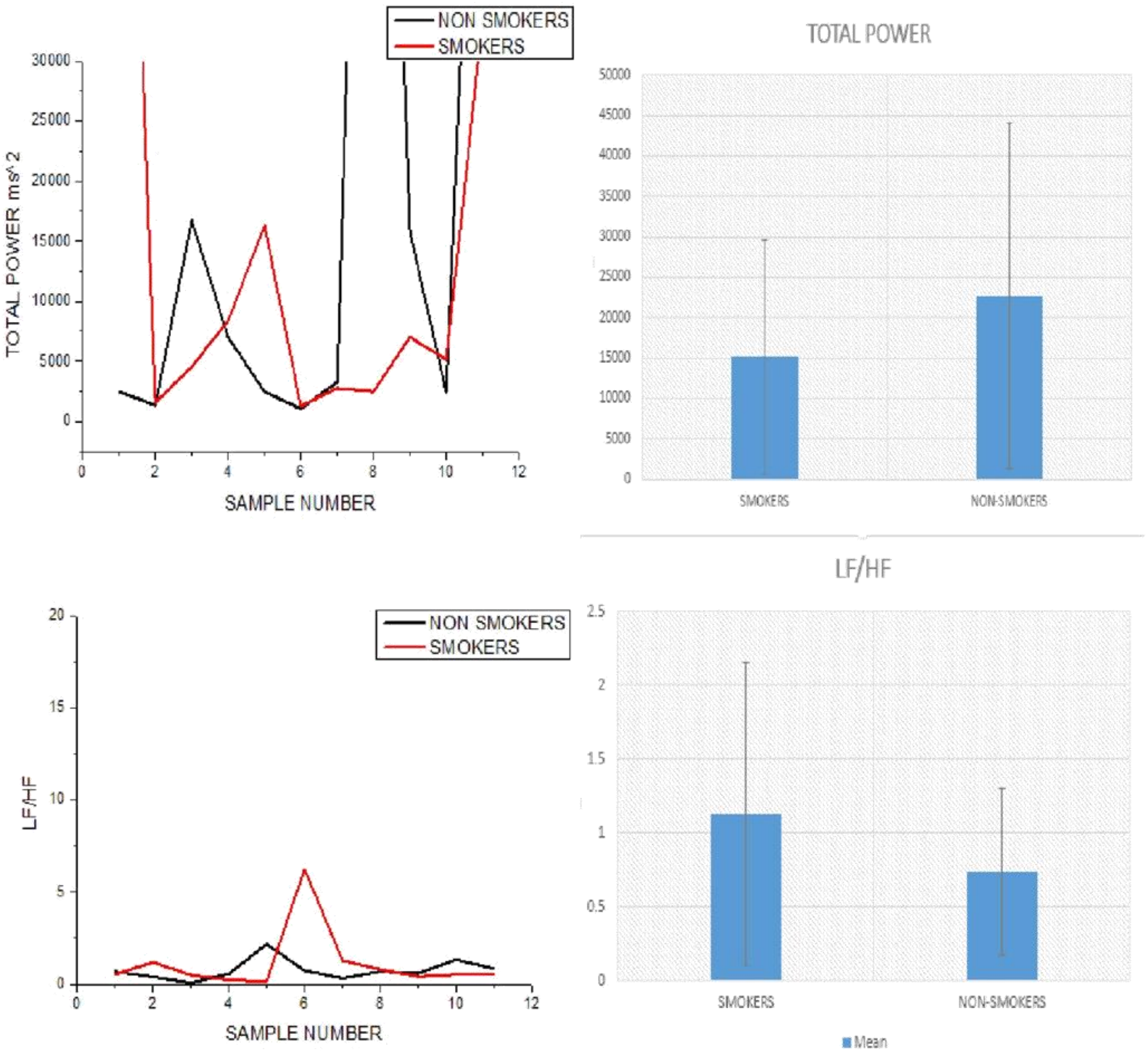


Figure 16: Graphs plotted Frequency domain indices.

TABLE 3. FREQUENCY DOMAIN ANALYSIS

CATEGORY	VLF (Hz)	LF (Hz)	HF (Hz)	VLF (ms^2)	LF (ms^2)	HF (ms^2)	Total power (ms^2)
NON-SMOKERS							
Mean	0.0184	0.09	2.73	3847.3	7910.89	8992.2	22674.46
SD	0.01	0.03	0.07	1453.341	987.484	4853.5	1346.164
SMOKERS							
Mean	0.012429	0.10831	0.2294	1718.688	4376.575	8992.186	15119.24
SD	0.0106	0.0259	0.040145	814.682	895.72	4853.55	4472.16

4.1.3 NONLINEAR RESULTS

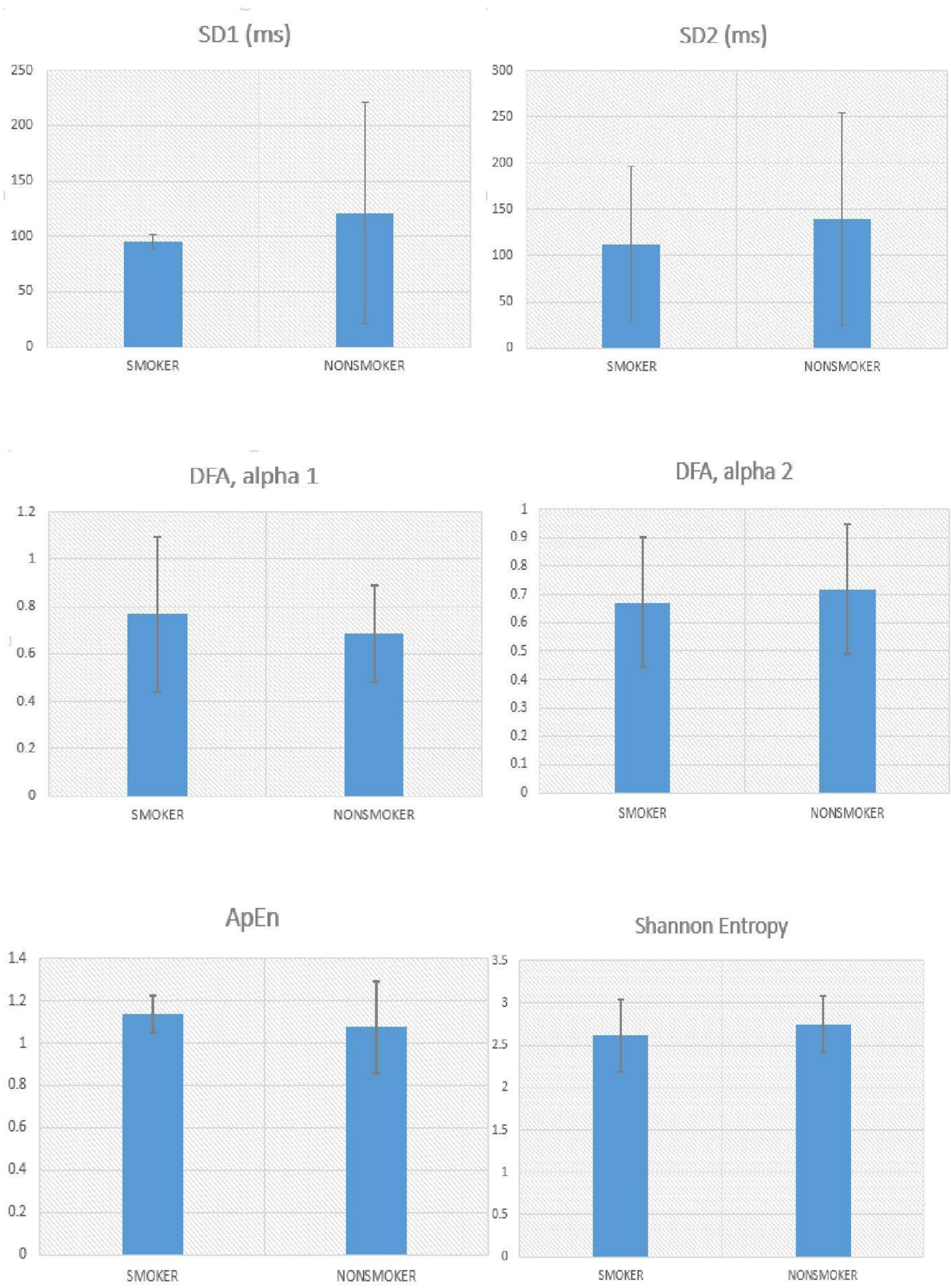


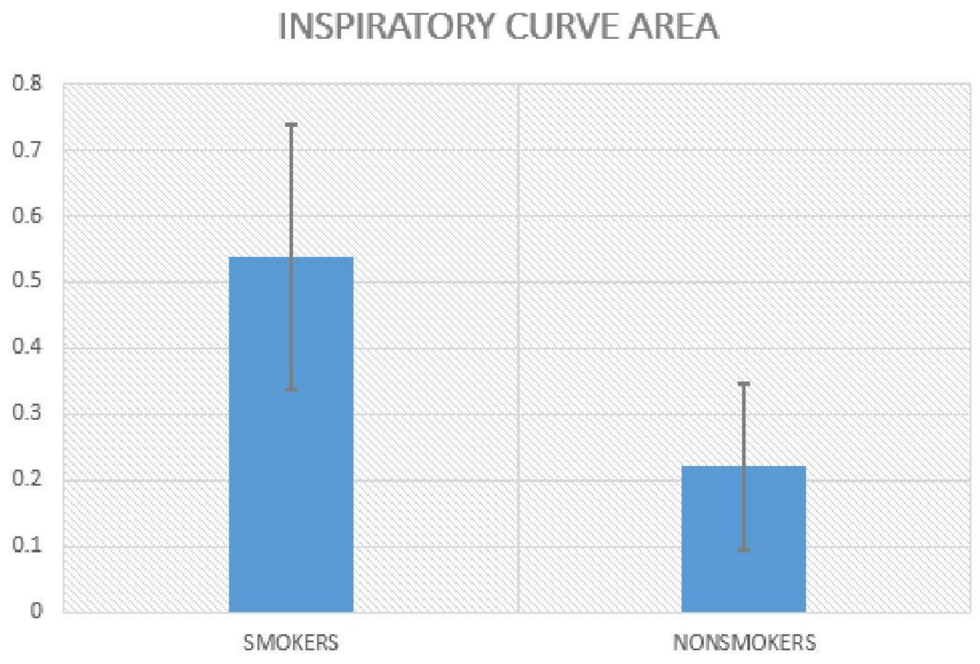
Figure 17: Graphs plotted for Nonlinear- indices.

Table 4: Data obtained from Non-linear analysis.

NONLINEAR ANALYSIS

Categogy	Poincare plot SD1(ms)	Poincare plot SD2(ms)	Detrended fluctuation analysis (DFA) ALPHA1	Detrended fluctuation analysis (DFA) ALPHA2	Approximate entropy, ApEn
NON-SMOKERS					
Mean	121.0073	139.65	0.6856	0.718	1.075
SD	99.8	114.67	0.203	0.229	0.216
SMOKERS					
Mean	95.09471	112.179	0.76801	0.67201	1.1348
SD	6.159003	83.8	0.3278	0.23	0.0867

4.2 RESPIRATORY SIGNAL ANALYSIS



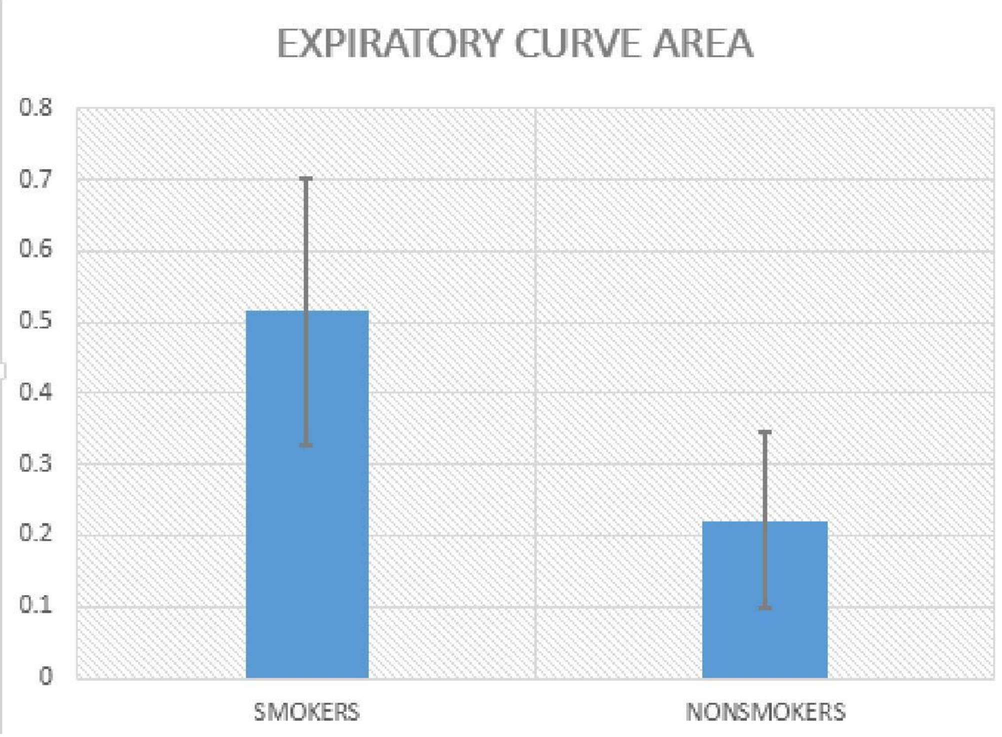


Figure 18: Graphs for inspiratory and expiratory areas with standard deviation. .

4.3 DISCUSSIONS

From the above results, it can be observed the mean heart rate of the smokers was less than the non-smokers whereas the mean R-R interval was more in case of smokers. From the frequency domain analysis it was observed that there was significant increase in the low frequency (LF) indices in case of smokers and slight increase in the high frequency (HF) indices for non-smokers. Thus, the LF/HF ratio was higher in case of smokers. From this we can deduce, that the sympathetic activity in case of smokers was high as compared to the non-smokers. The power spectral analysis in frequency domain also have significant difference between smokers and non-smokers. From the analysis of the nonlinear indices we observed Shannon entropy, Detrended Fluctuation Analysis (DFA) and Poincare plot parameters (PPA). Here, the Shannon entropy was the quantity that quantifies the repetition pattern in a signal, DFA refers to the analysis of the series of R-R intervals and the Poincaré HRV plot is a graph in which each R-R interval is plotted against next R-R interval with two standard descriptors. The Shannon entropy was found to be more or less similar for both smokers and non-smokers. The Poincare standard descriptors seem to be less for smokers in comparison with the non-smokers.

From the analysis of the respiratory signals, it was observed that both the inspiratory and expiratory areas under the respective curves were higher for the smokers in comparison to the non-smokers.

5. CONCLUSION

The results were analysed and it can be concluded that the mean heart rate of the smokers was less than the non-smokers whereas the mean R-R interval was more in case of smokers. There was significant increase in the low frequency (LF) indices for smokers and there was increase in the high frequency (HF) indices for non-smokers. The LF/HF ratio was thus higher in case of smokers. The power spectral analysis in frequency domain showed significant difference between smokers and non-smokers. The Shannon entropy analysed from the nonlinear indices of HRV was found to be more or less similar for both smokers and non-smokers. The Poincare standard descriptors seem to be less for smokers in comparison with the non-smokers. The analysis of the respiratory signals concluded that both the inspiratory and expiratory areas under the respective curves were higher for the smokers in comparison to the non-smokers. From all these we can deduce that the sympathetic activity in case of smokers is high as compared to the non-smokers. Thus, the overall study concluded that the integrated data acquisition system is a good methodology to investigate both ECG and respiratory signals. In future, this methodology can be used to find the relation between R-R intervals and breathing intervals to investigate the co-relative function between heart and lungs in various pathological cases.

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ANNEXURE-I

Questionnaire

Instructions:

1. The questionnaire is being conducted for a research thesis.
2. Fill in the information honestly and to the best of your knowledge.
3. The information you have provided will be used only for the purposes of this project, and will be kept strictly confidential.

Section I. General Information

1.1 Name: 1.2. D.O.B:

1.3 Age: 1.4 Sex: M / F:

1.8 Contact address:

1.9 Email id:

1.10 Do you have any objection or difficulty in giving this information to us? Yes/ No

1.11 Do you smoke? Yes/ No. If yes, how frequently.

1.12 Consumption of alcohol? Yes/ No. If yes, how frequently.

1.13 Please state medical history of self and family (If any):

1.14 Do you involve yourself in any physical activity (gym, outdoor sports, cycling etc.). If yes, please specify the frequency and activity.

Declaration:

I___________________________________, hereby give my consent to abide by the terms and conditions of the research experiment and understand that it is being conducted under my sole discretion.

Signature of the participant
Date:
Place:

Person in charge:
Santosh Kumar Sahoo
email id: 111bm0545@nitrkl.ac.in

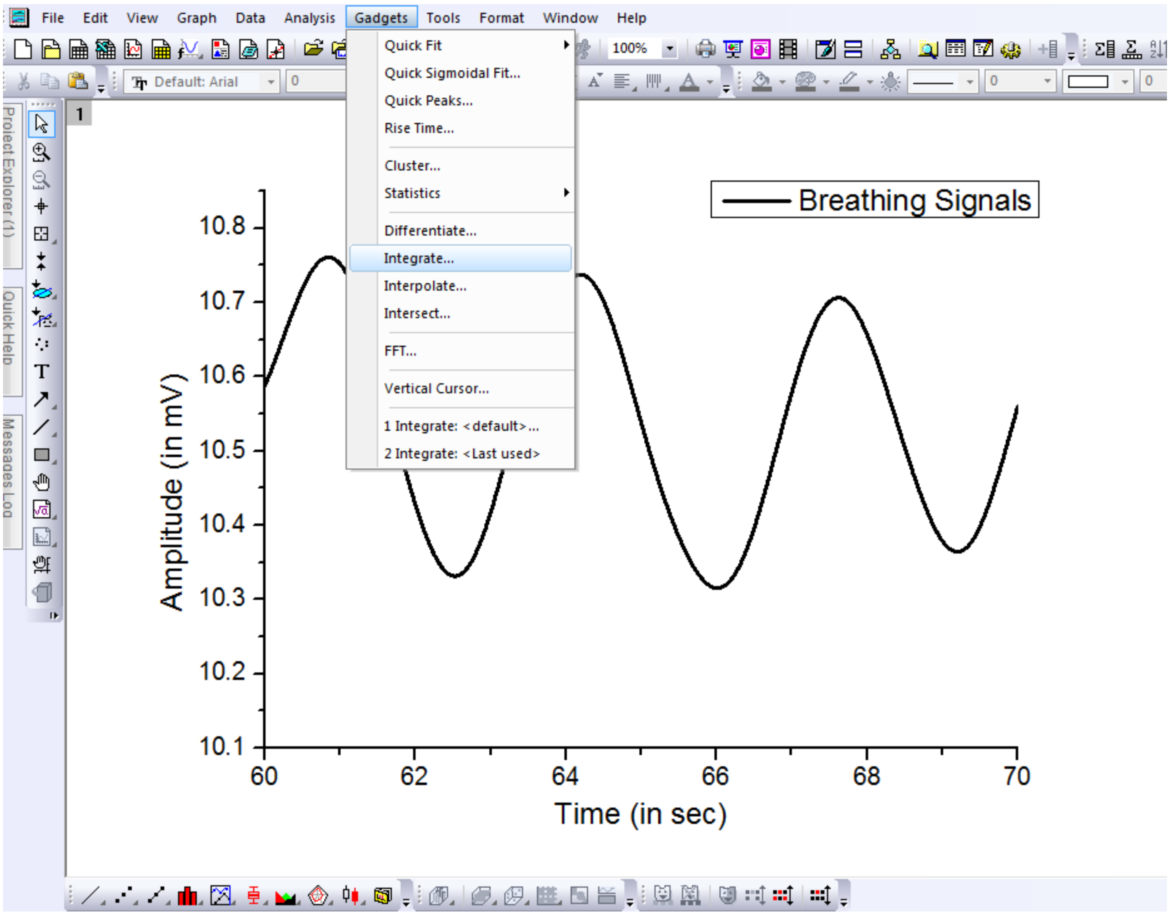
ANNEXURE-II

Procedure for finding the Inspiratory and Expiratory area (area under the curve)

The inspiratory and the expiratory areas are determined manually for each respiratory cycle for all the samples collected. This is done by using Origin 9 Pro (original version). The steps for finding the areas are mentioned below:

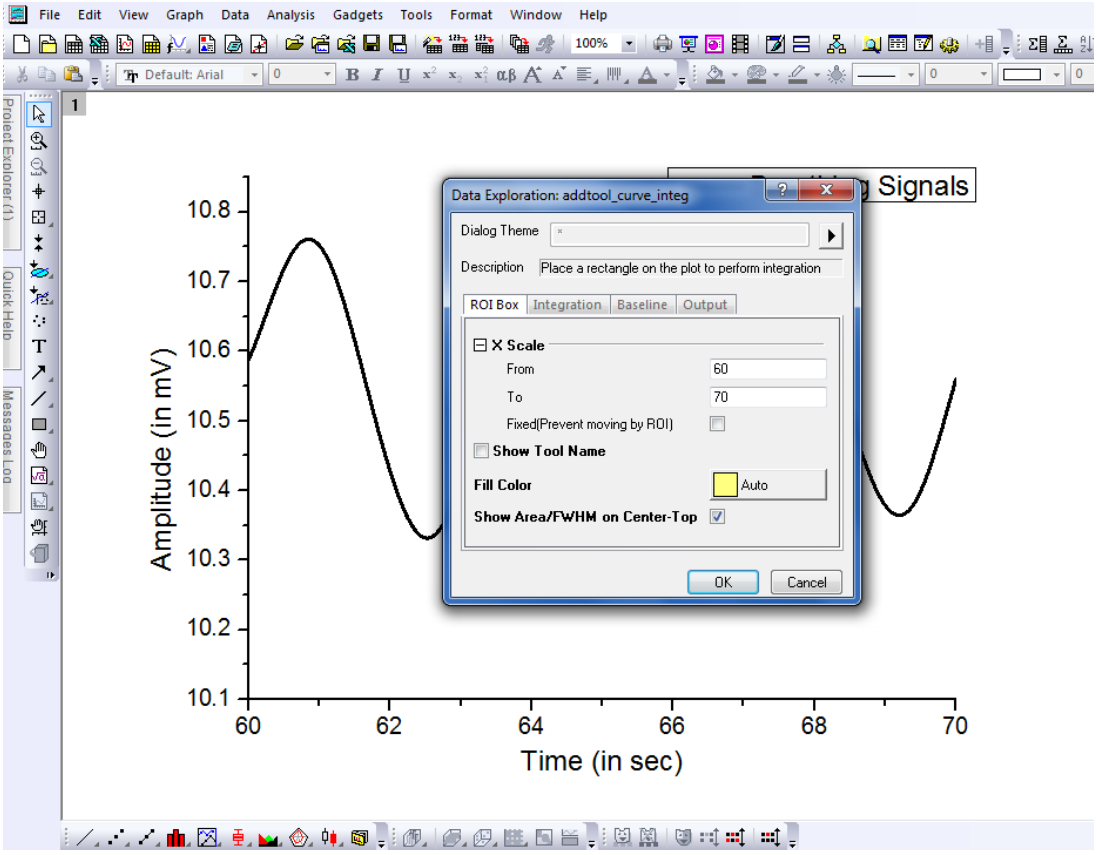
Step 1: The data were plotted in Origin 9 Pro and the scale was adjusted.

Step 2: Then from the 'Gadgets' section on the tool bar, the 'Integrate' tool was selected.

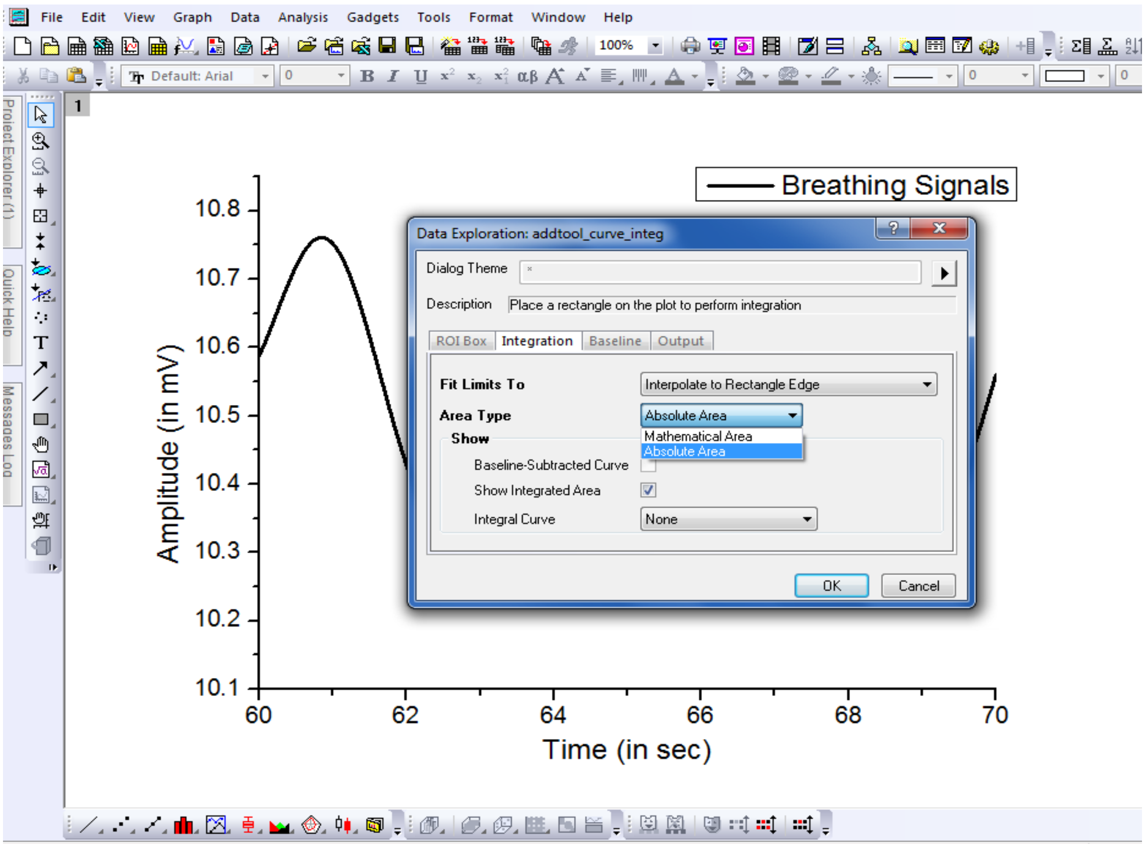


Step 3: After selecting the 'Integrate' tool, the tool box appears. Here we have four sections which have to select the parameters according to the data.

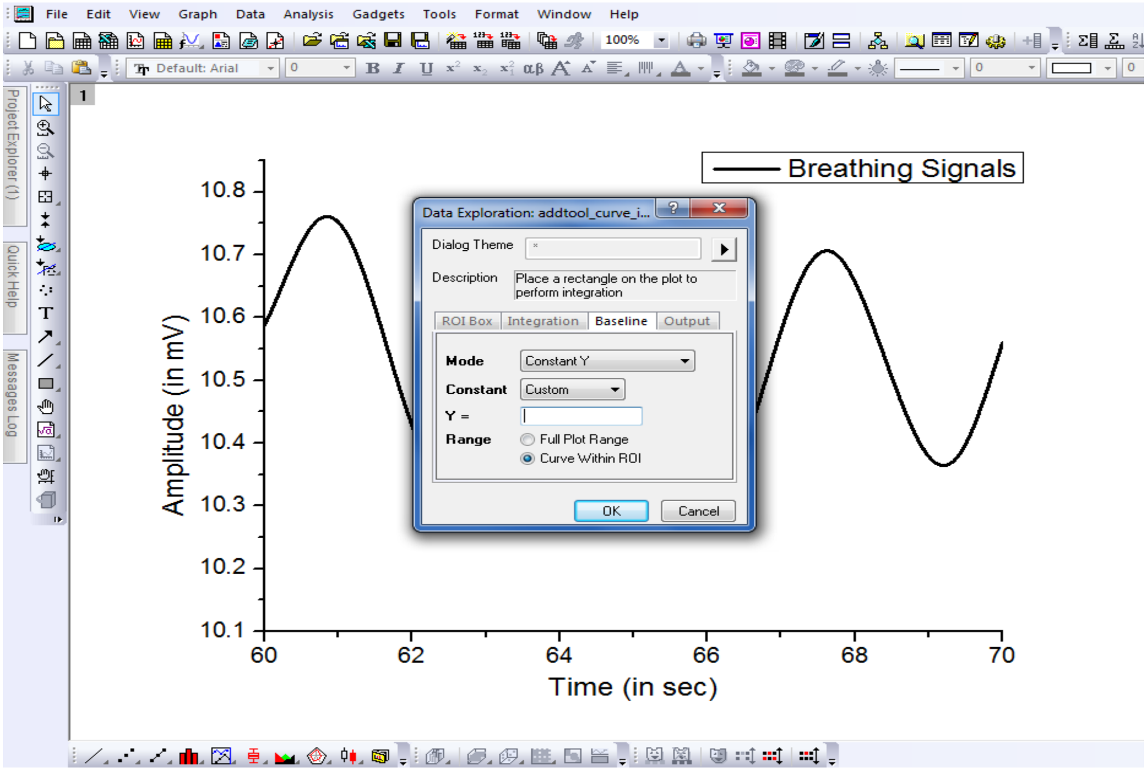
- In the ROI box section we have to select the range of plot to be integrated.



- In the ‘integration section the ‘absolute area ‘ was selected under the ‘Area Type ‘.



- In the ‘baseline’ section we have to select a constant Y and give the value of the minima for respective cycles.



- Under the ‘output’ various parameters were selected as per the requirement.
- This was repeated for both inspiration and expiration curves.